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Hanebuchi

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(54) **HAND-HELD OPHTHALMOLOGICAL
DEVICE**

(56) **References Cited**

U.S. PATENT DOCUMENTS

(75) Inventor: **Masaaki Hanebuchi**, Aichi (JP)
(73) Assignee: **NIDEK CO., LTD.**, Gamagori-shi (JP)
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5,528,323 A *	6/1996	Fujieda	A61B 3/103 351/211
5,646,709 A *	7/1997	Carter	A61B 3/145 351/205
5,859,687 A *	1/1999	Heine	A61B 3/1208 351/216
6,630,951 B1	10/2003	Suzuki		
2008/0309877 A1	12/2008	Saito et al.		

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FOREIGN PATENT DOCUMENTS

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JP	A-6-46999	2/1994
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JP	A-2000-69353	3/2000
JP	A-2001-161644	6/2001

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OTHER PUBLICATIONS

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Primary Examiner — Hung Dang

(74) *Attorney, Agent, or Firm* — Oliff PLC

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A61B 3/00	(2006.01)
A61B 3/14	(2006.01)

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(2013.01); **A61B 3/103** (2013.01); **A61B**
3/1035 (2013.01); **A61B 3/14** (2013.01)

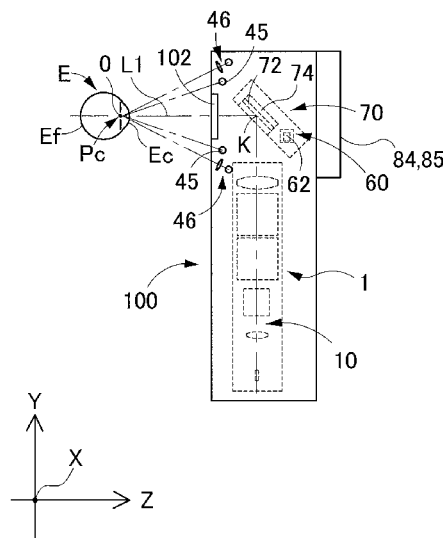
(58) **Field of Classification Search**

CPC A61B 3/1208
USPC 351/218, 216, 221, 211, 246
See application file for complete search history.

(57) **ABSTRACT**

A hand-held ophthalmological device includes: a main unit having an ophthalmoscopic optical system configured to project ophthalmoscopic light to an examinee's eye and receive reflection light therefrom to examine or measure an examinee's eye; a detector placed in the main unit and configured to detect a relative deviation between an optical axis of the ophthalmoscopic optical system and the examinee's eye; a deviation compensating optical system placed as a part of the ophthalmoscopic optical system and configured to compensate the deviation; and a drive part configured to drive the deviation compensating optical system based on output from the detector.

20 Claims, 9 Drawing Sheets



(56)

References Cited

FOREIGN PATENT DOCUMENTS

JP 2003-189221 A 7/2003
JP 2006-071743 A 3/2006

JP A-2008-197323 8/2008
JP A-2008-307104 12/2008
JP A-2011-164321 8/2011
JP A-2012-10952 1/2012

* cited by examiner

FIG. 1

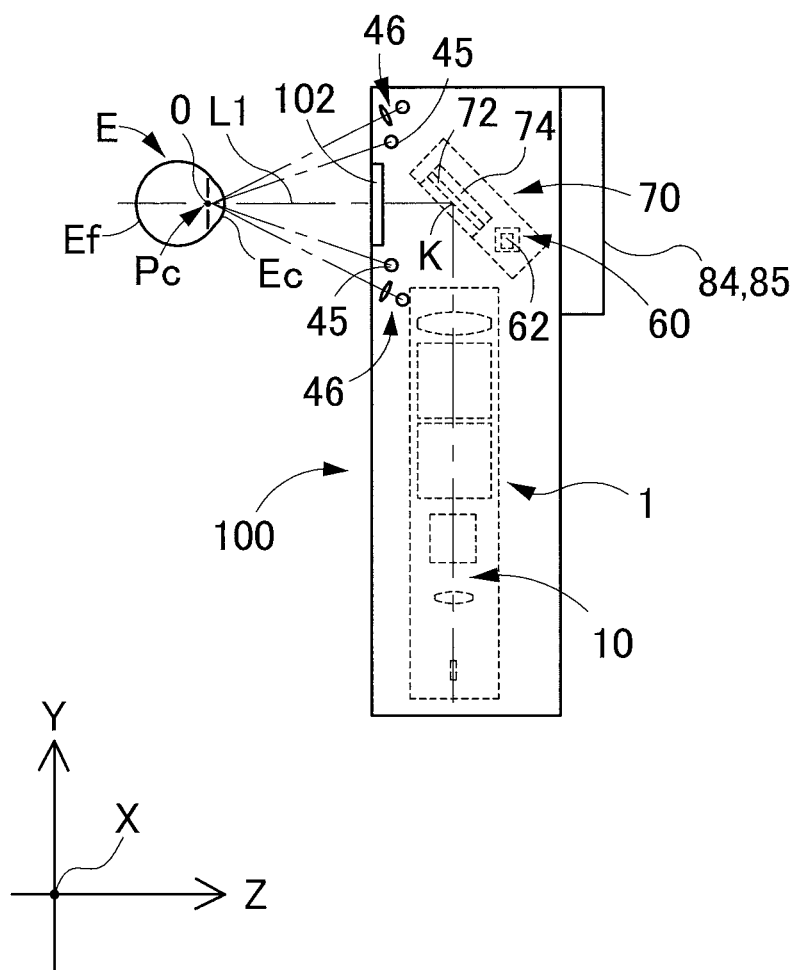


FIG. 2

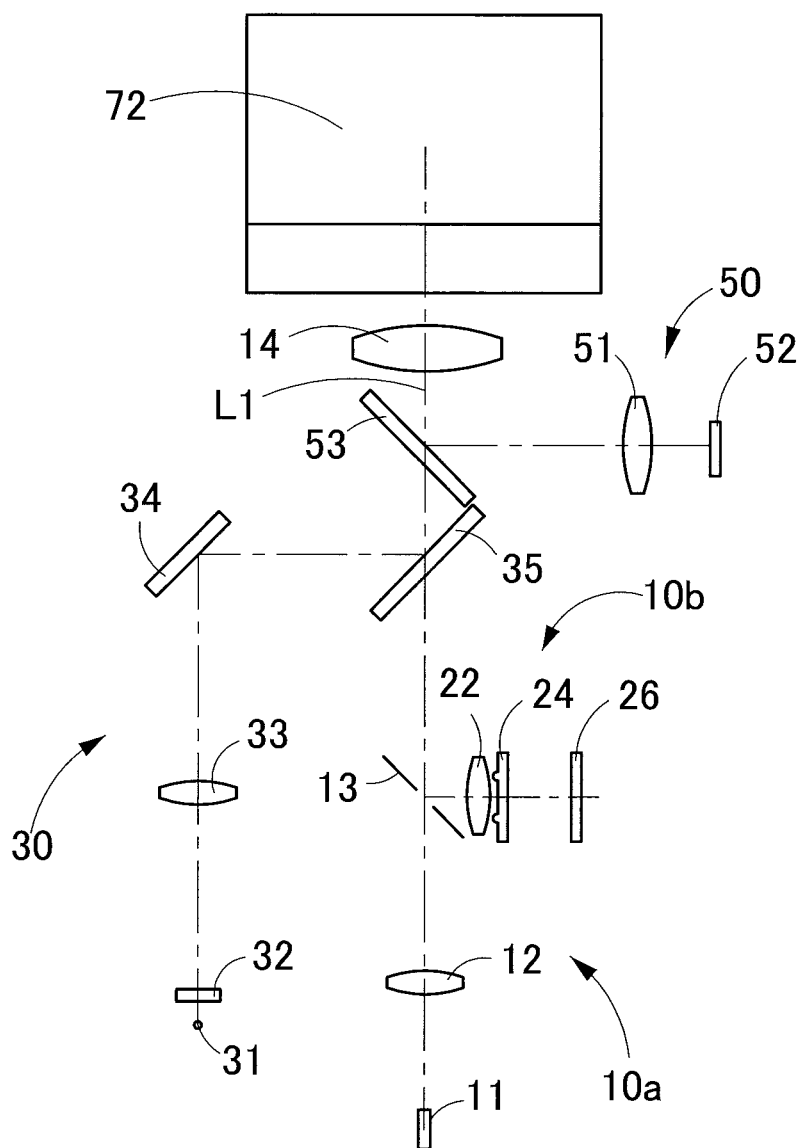


FIG. 3

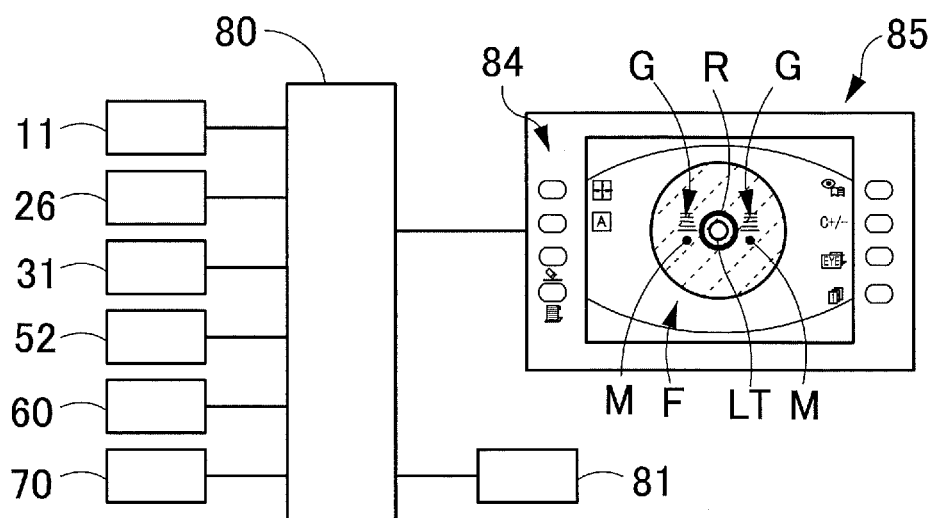


FIG. 4

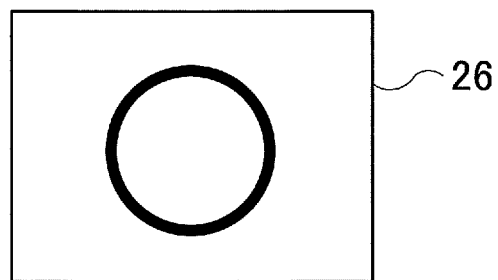


FIG. 5 (a)

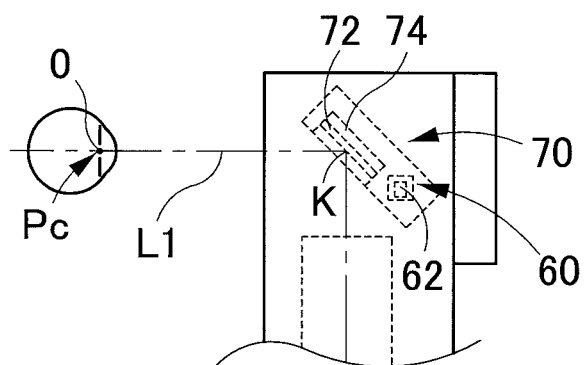


FIG. 5 (b)

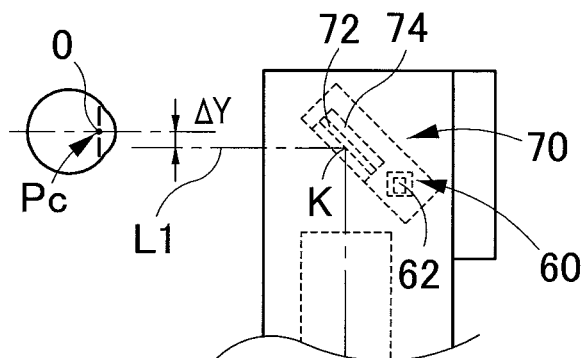


FIG. 5 (c)

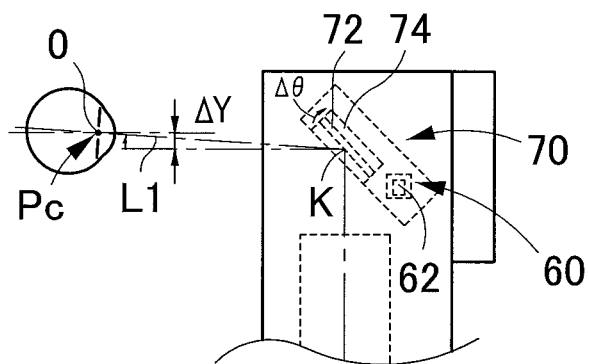


FIG. 6

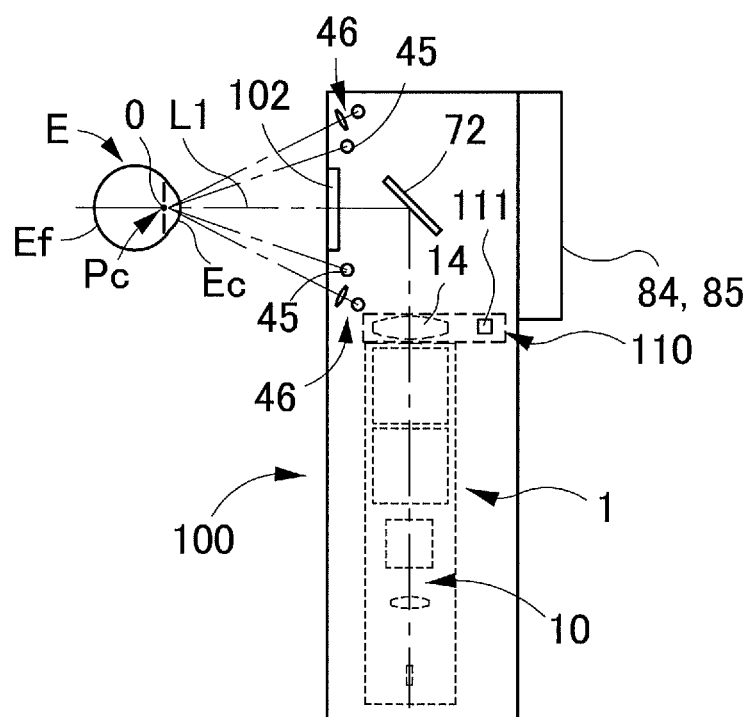


FIG. 7

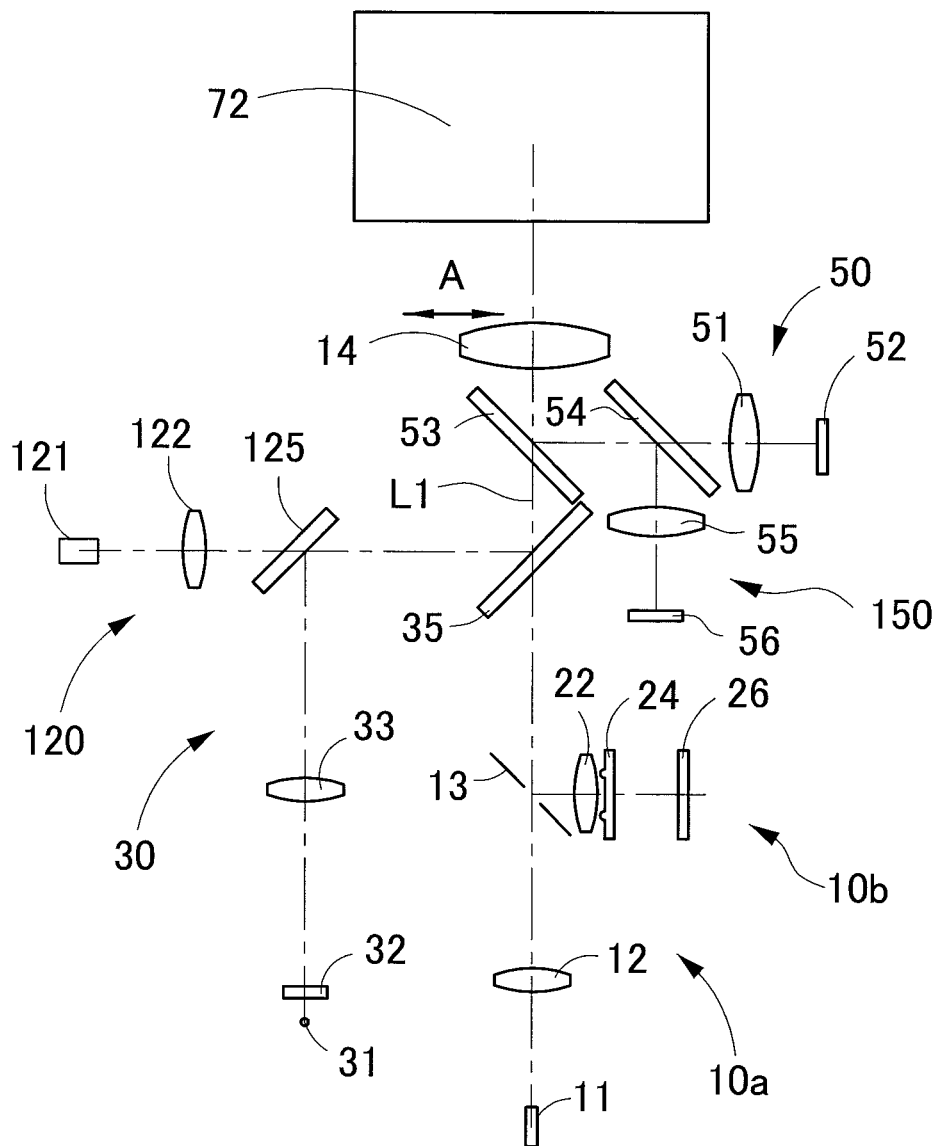


FIG. 8

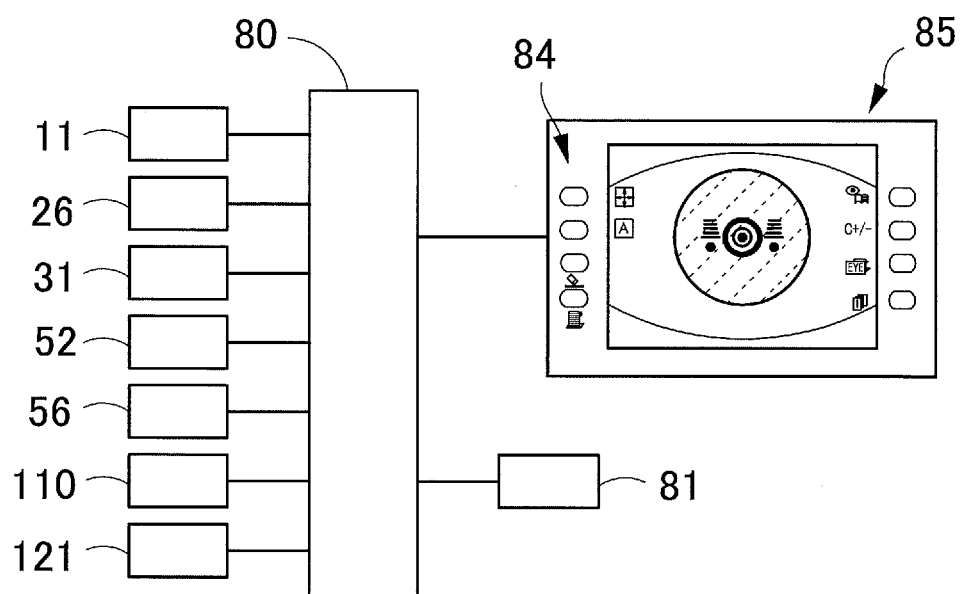


FIG. 9

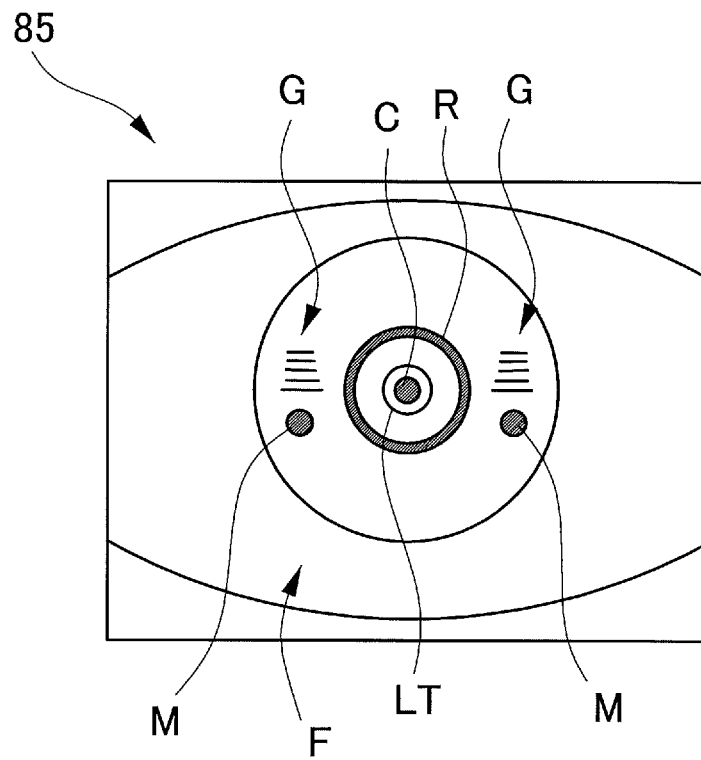


FIG. 10 (a)

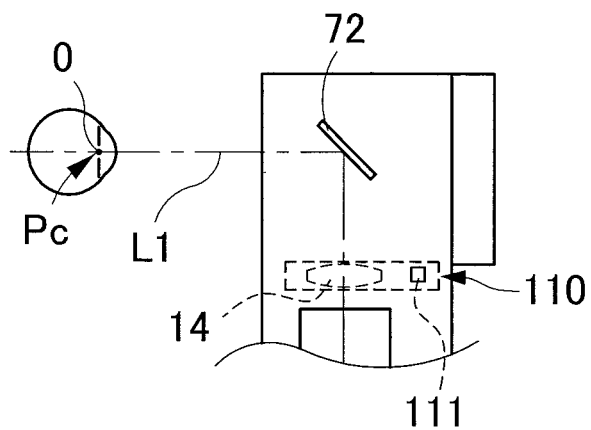


FIG. 10 (b)

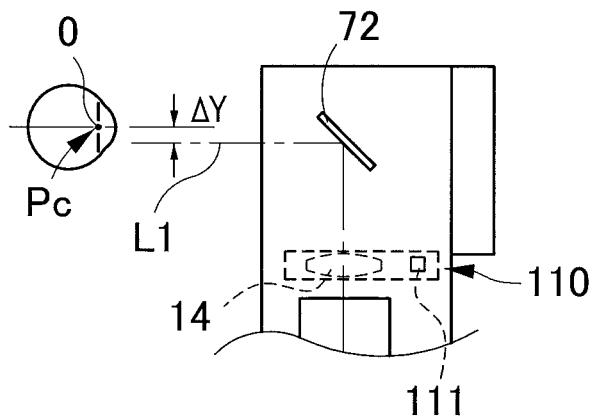
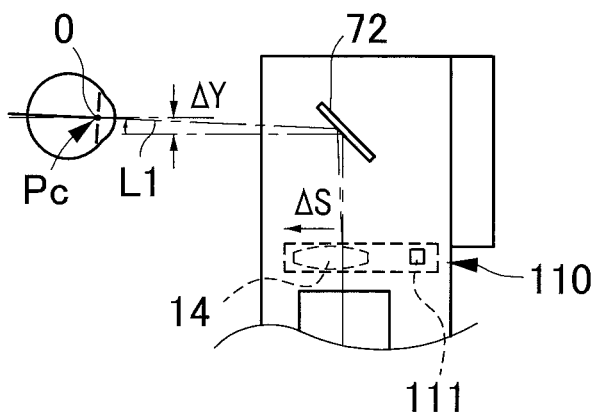


FIG. 10 (c)



**HAND-HELD OPHTHALMOLOGICAL
DEVICE****CROSS-REFERENCE TO RELATED
APPLICATIONS**

This is a national phase application based on the PCT International Patent Application No. PCT/JP2011/080352 filed on Dec. 27, 2011, the entire contents of which are incorporated herein by reference.

TECHNICAL FIELD

The present invention relates to a hand-held ophthalmological device for performing examination and measurement of an eye.

BACKGROUND ART

As an ophthalmological device for measuring and examining eyes, there are known an auto-refractometer and a fundus camera, for example. These devices include a floor-standing type device installed in an examination room and a portable hand-held type device (see Patent Document 1).

In the case of the floor-standing type device, this device is wholly moved in three dimensions by an operation member such as a joystick. Such a device is heavy and stable and thus provides relatively high operability for alignment. However, this device is hard to move.

RELATED ART DOCUMENTS**Patent Documents**

Patent Document 1: JP-A-11(1999)-19039

SUMMARY

On the other hand, in the case of the hand-held type device, an examiner holds the whole device and moves it three-dimensionally. This device may be moved with respect to an eye due to hand shake or minute movement of the examiner. From this viewpoint, it is conventionally difficult to make alignment of the hand-held type ophthalmological device, which is apt to cause measurement/examination errors.

The present invention has been made to solve the above problems and has a purpose to provide a hand-held ophthalmological device capable of providing high operability and performing stable measurement and examination.

To achieve the above purpose, one aspect of the invention provides a hand-held ophthalmological device includes: a main unit having an ophthalmoscopic optical system configured to project ophthalmoscopic light to an examinee's eye and receive reflection light therefrom to examine or measure an examinee's eye; a detector placed in the main unit and configured to detect a relative deviation between an optical axis of the ophthalmoscopic optical system and the examinee's eye; a deviation compensating optical system placed as a part of the ophthalmoscopic optical system and configured to compensate the deviation; and a drive part configured to drive the deviation compensating optical system based on output from the detector.

EFFECTS OF THE INVENTION

The invention can provide high operability and perform stable measurement and examination.

MODE FOR CARRYING OUT THE INVENTION**Summary**

5 An ophthalmological device in a present embodiment is a hand-held ophthalmological device including a main unit, a detector to detect a displacement, a displacement compensating optical system, and a drive part to drive the displacement compensating optical system.

10 The main unit has an ophthalmoscopic optical system. This ophthalmoscopic optical system is provided to project ophthalmoscopic light to an eye E and receive reflection light therefrom to examine or measure the eye E. The ophthalmoscopic optical system is an ophthalmoscopic optical system configured for example to project a measurement index to an eye to be examined and receive reflection light therefrom through a light receiving element to thereby measure ophthalmic optical characteristics of the eye. Typical examples of such a device are an auto-refractometer and a keratometer. 20 The ophthalmoscopic optical system may be a photographing optical system (e.g., a fundus camera, an ophthalmic OCT) to photograph an image of an eye.

The detector to detect a displacement is placed in the main unit and provided to detect a relative positional displacement between an optical axis of the ophthalmoscopic optical system and the eye E. In this case, it is more effective that the detector is configured to optically detect the relative positional displacement between the optical axis of the ophthalmoscopic optical system and an anterior segment of the eye. 30 For instance, there are provided a projection optical system configured to form an index image at a position corresponding to a nearly corneal apex of the eye E and an optical position sensor is provided to detect the position of reflection light of the index image. Based on a light reception signal from the optical position sensor, a relative displacement between the optical axis of the ophthalmoscopic optical system and the eye E in a direction perpendicular to the optical axis of the ophthalmoscopic optical system is detected at high speed. 40

An effective one as the detector is, for example, a deviation detector to detect a relative deviation (first displacement) between the optical axis of the ophthalmoscopic optical system and the eye E. This deviation includes a deviation applied to the main unit (shake or minute movement of an examiner's hand) and a deviation caused by the movement of an examinee. Conceivable configurations as the detector are for example an angular velocity sensor, an acceleration sensor, and others configured to detect a deviation applied to the main unit, and also a light receiving sensor (e.g., an optical position sensor, an imaging element) configured to receive reflection light from an eye (e.g., a cornea), an ultrasonic sensor, and others. In this case, an effective one is a shift deviation detector to detect a shift deviation applied to the main unit. 50

The aforementioned detector may be configured to detect a displacement (second displacement) caused when the eye E is displaced from the optical axis of the ophthalmoscopic optical system. As this detector, for example, there are conceived a light receiving sensor (e.g., an optical position sensor, an imaging element) configured to receive reflection light from an eye (e.g., a cornea), an ultrasonic sensor, and others. 60

The aforementioned detector may also be configured to detect both the first displacement and the second displacement. As this detector, for example, there are conceived a light receiving sensor (e.g., an optical position sensor, an imaging element) configured to receive reflection light from an eye, an ultrasonic sensor, and others. 65

The compensating optical system is placed as a part of the ophthalmoscopic optical system and configured to compensate or correct a relative positional displacement between the optical axis of the ophthalmoscopic optical system and the eye E. This compensating optical system is used to compensate a detected positional displacement based on output from the aforementioned detector. It is effective that the deviation compensating optical system is configured to compensate a relative deviation between the optical axis of the ophthalmoscopic optical system and the eye E (first displacement). Further, the deviation compensating optical system is configured to compensate a displacement (second displacement) caused when the eye E is displaced from the optical axis of the ophthalmoscopic optical system. The deviation compensating optical system may be configured to compensate both the first displacement and the second displacement.

As the compensating optical system, for example, a light deflection member to deflect ophthalmoscopic light is used. It is effective that the deviation compensating optical system is configured to allow a lens placed in an optical path of the ophthalmoscopic optical system to move in a vertical direction to the lens optical axis. In this case, for instance, an objective lens is moved. Furthermore, the deviation compensating optical system may be configured to place a light reflection member movably in the optical path of the ophthalmoscopic optical system.

For instance, the compensating optical system is placed in a common optical path of a light projecting system and a light receiving system of the ophthalmoscopic optical system or placed in each of the light projecting system and the light receiving system of the ophthalmoscopic optical system. In a case where an imaging optical system is provided to capture a front image of an examinee's eye through an imaging element, effectively, this optical system is placed in a common optical path of the ophthalmoscopic optical system and the imaging optical system, for example.

The compensating optical system may be configured to be placed in an optical path of one of the light projecting system and the light receiving system of the ophthalmoscopic optical system. In the case of the keratometer, for example, the compensating optical system may be placed only in the light receiving system (e.g., an anterior segment observation system) configured to receive light of an index image. In a configuration capable of measuring both eye refractive power and corneal shape, the compensating optical system may be placed in one of an eye refractive power measuring optical system and a corneal shape measuring optical system.

The aforementioned drive part is provided to drive the compensating optical system based on output from the aforementioned detector. The drive part is for example an actuator capable of moving a lens in a vertical direction relative to an optical axis of the lens and an actuator capable of rotating a mirror. Further, a controller (e.g., a CPU, a control circuit) provided in the main unit controls the drive part based on output from the detector.

In a case where the drive part is provided in the eye refractive power measuring device, effectively, this drive part is configured to drive the compensating optical system so as to bring a region to be measured by the eye refractive power measuring optical system onto a pupil of the eye E.

Further effectively, a computing unit is provided to compensate an eye examination result obtained by the ophthalmoscopic optical system according to the position of the optical member provided in the deviation compensating optical system. For instance, the computing unit provided in the device compensates a measurement result of eye optical char-

acteristics of the eye E according to the position of the lens provided in the compensating optical system.

According to the ophthalmological device in the present embodiment can provide high operability and perform stable measurement/examination. In the case of the eye characteristic measuring device to measure eye optical characteristics, for example, difficulty in positional alignment due to hand shake can be solved during measurement by the eye characteristic measuring device.

In the case of the eye photographing device, when a displacement between the imaging optical axis and the eye E is detected, the compensating optical system is driven based on a detection signal. In the case of a device for imaging a fundus, for example, a control part **80** may be configured to detect a deviation of a fundus image captured by the imaging element for fundus imaging (e.g., for alignment) provided in a fundus observation optical system, and then feedback drive the compensating optical system to compensate the deviation. Of course, an accelerator sensor or the like may be provided.

The present embodiment is applicable to not only the hand-held ophthalmological device but also a floor-standing ophthalmological device in case it is installed in an unstable place.

In the present embodiment, furthermore, a light deflection member (e.g., a prism) is provided in a position off a pupil conjugation position and this member is eccentrically rotated about the optical axis, thereby appropriately avoiding disturbance light from entering in an ophthalmoscopic light receiving element.

EXAMPLES

First Example

A first example of the present embodiment will be explained below. The first example of the invention is explained referring to the accompanying drawings. FIG. 1 is an external side view showing one example of a hand-held ophthalmological device of the first embodiment. The following explanation exemplifies an eye refractive power measuring device (an auto-refractometer). In the following explanation, a positional relationship between an examinee's eye and the device is defined assuming that a back and forth direction is a Z direction, a right and left direction is an X direction, and an up and down direction is a Y direction while the examinee's eye and the device face each other.

A main unit **100** is provided with an optical unit **1** including an ophthalmoscopic optical system **10** to project an ophthalmoscopic light beam to the examinee's eye E and receiving reflection light therefrom, a deviation detection unit **60** to detect a deviation (e.g., hand shake of an examiner) applied to the main unit **100**, and a deviation compensating unit **70** to compensate a deviation of the ophthalmoscopic light caused by the deviation applied to the main unit **100**. A test window **102** is positioned in a side of the main unit **100** that will face the eye E. An operation part **84** and a monitor **85** are positioned in a side of the main unit **100** that will face an examiner. Furthermore, the main unit **100** contains an electric system for control and arithmetic processing. The examiner faces the examinee and holds the main unit **100**. While looking at the monitor **85**, the examiner makes alignment of the main unit **100** with respect to the eye E.

The deviation compensating unit **70** is provided with a reflection mirror **72** serving as a deviation compensating optical system, and a drive part **74** (e.g., a voice coil motor mechanism) to rotate the reflection mirror **72** in the X and Y directions. The reflection mirror **72** is placed in one of the

optical paths of the ophthalmoscopic optical system 10 and is used as a part of the ophthalmoscopic optical system 10. As an alternative, a plurality of reflection mirrors for deviation compensation may be provided (e.g., utilizing a galvanometer mechanism). In this case, one is rotated in the X direction and the other is rotated in the Y direction.

In FIG. 1, the reflection mirror 72 is placed in a position facing the eye E and the ophthalmoscopic optical system 10 is placed below the reflection mirror 72. An optical axis L1 of the ophthalmoscopic optical system 10 is deflected toward the eye E by the reflection mirror 72. The reflection mirror 72 reflects light emitted from the ophthalmoscopic optical system 10 toward the eye E and reflects reflection light from the eye E into the ophthalmoscopic optical system 10.

The reflection mirror 72 is rotated about an intersection point K between the optical axis L1 and the reflection mirror 72. A reference position of the reflection mirror 72 is a position in which the optical axis L1 is parallel to the Z direction. As the reflection mirror 72 is rotated, the optical axis L1 pivots about a predetermined point O on the optical axis L1 (see FIGS. 5(a) to 5(c)). The predetermined point O is brought into coincidence with a pupil center Pc by alignment.

The detecting unit 60 includes at least one of an angular velocity sensor and an accelerator sensor to detect a deviated state of the main unit 100. The detecting unit 60 is connected to the drive part 74 which is connected to the reflection mirror 72. With such configuration, the deviation detecting unit 60 detects a deviation amount and the drive part 74 drives the reflection mirror 72 based on output of the deviation detecting unit 60.

In FIG. 1, the detecting unit 60 includes an accelerator sensor 62. This sensor 62 outputs an accelerator signal according to a parallel deviation (shift deviation) of the main unit 100 in the X and Y directions. In addition to the X-Y two-axis sensor, a sensor configured to detect a parallel deviation in the Z direction may also be provided. As an alternative, a single-axis sensor may be provided alone. As the accelerator sensor 62, there are utilizable mechanical, optical, and semiconductor-type (capacitance type, piezoresistance type, gas temperature distribution type) accelerator sensors. The reason why the accelerator sensor 62 is used is that a distance between the main unit 100 and the eye E in the eye examination and measurement is relatively short and thus the parallel deviation is liable to be caused by the examiner. In addition to the accelerator sensor 62, an angular velocity sensor may be provided. In this case, an angular velocity signal according to a rotation deviation of the main unit 100 is output.

FIG. 2 is an optical arrangement diagram showing optical systems contained in the main unit 100 seen from front in the first embodiment. The optical unit 1 is provided with the ophthalmoscopic optical system 10, a light projecting optical system 30 to project a fixation target to the eye E, and an observation optical system 50 to allow observation of the eye E.

In FIG. 2, the ophthalmoscopic optical system 10 is an optical system to objectively measure eye refractive power. The ophthalmoscopic optical system 10 projects a measurement index to a fundus Ef and receives reflection light from the fundus. Based on the light receiving signal, the refractive power of the eye E is measured.

More concretely, the ophthalmoscopic optical system 10 consists of a light projecting optical system 10a to project a spot index onto the fundus Ef through a pupil center Pc and a light receiving optical system 10b to extract fundus reflection light as a ring image through the periphery of a pupil and form a ring-shaped reflection image on an imaging element 26.

The light projecting optical system 10a includes a light source 11, a relay lens 12, a hole mirror 13, an objective lens 14, and the reflection mirror 72, which are arranged on the optical axis L1. The light source 11 is placed in a nearly conjugated position with the fundus Ef and an aperture of the hole mirror 13 is placed in a nearly conjugated position with the pupil of the eye E.

Measurement light emitted from the light source 11 is reflected by the reflection mirror 72 via the relay lens 12, the hole mirror 13, a dichroic mirror 35, a dichroic mirror 53, and the objective lens 14. The measurement light reflected by the reflection mirror 72 passes through the pupil center Pc, projecting a spot-shaped beam onto the fundus Ef.

The light receiving optical system 10b includes the reflection mirror 72, the objective lens 14, the hole mirror 13, a collimator lens 22, a ring lens 24, and the imaging element (e.g., a two-dimensional imaging element such as a CCD and a CMOS) 26. The imaging element 26 is placed in a nearly conjugated position with the fundus Ef via the lens 14, the lens 22, and the ring lens 24. The ring lens 24 consists of a lens part formed with an annular, cylindrical lens and a light shielding part having a ring aperture equal in size to the lens part, and is placed in a nearly conjugated position with the pupil of the eye E. An output signal from the imaging element 26 is connected to the control part 80.

The fundus reflection light by the light projecting optical system 10a is reflected again by the reflection mirror 72, and reflected again by a reflection plane of the hole mirror 13 via the objective lens 14, the dichroic mirror 53, and the dichroic mirror 35, and then is collimated into a nearly parallel beam (in a case of an emmetropic eye) by the collimator lens 22. The ring-shaped beam is extracted by the ring lens 24 and received as a ring image by the imaging element 26.

Instead of the above configurations, various methods may be adopted; e.g., a method of projecting a ring index onto the fundus Ef through the peripheral part of a pupil and extracting reflection light through the pupil center so that a ring image is received by the imaging element, a phase difference method performed by projecting slit light onto a fundus, and others.

Between the objective lens 14 and the hole mirror 13, the dichroic mirror 35 serving as a beam splitter that reflects visible light and transmits infrared light is placed to deliver the light from the light source 31 toward the eye E. Between the dichroic mirror 35 and the objective lens 14, the dichroic mirror 53 serving as a beam splitter is placed to deliver the light from an anterior segment to the observation optical system 50. The mirror 53 has wavelength characteristics that transmit measurement light and reflect observation light. By those beam splitters, the measurement optical axis, the fixation optical axis, and the observation optical axis are made coaxial with the same optical axis (L1).

The target projecting optical system 30 includes a light source 31 that emits visible light, a fixation target 32 on which a scenic/animal image or others is painted, a light projecting lens 33, a total reflection mirror 34, the dichroic mirror 35, the objective lens 14, and the reflection mirror 72. In addition to the above configuration, as a fixation target, a point light source such as LED and a display such as a liquid crystal display, and others are used. Furthermore, a plurality of fixation targets may be placed two-dimensionally.

The fixation target 32 illuminated by the light source 31 is projected onto the fundus Ef via the light projecting lens 33, the total reflection mirror 34, the dichroic mirror 35, the dichroic mirror 53, the objective lens 14, and the reflection mirror 72. Accordingly, the eye E is induced to hold fixation. The light source 31 and the fixation target 32 are moved in the optical axis direction to apply a fogging to the eye E.

On the outside of the test window **102**, first projection optical systems **45** to project ring-shaped finite index images onto the cornea **Ec** are arranged concentrically with the optical axis **L1**. Furthermore, second projection optical systems **46** to project infinite index images onto the cornea **Ec** are arranged left-right symmetric (up-down symmetric for convenience in the figure) with respect to the optical axis **L1**. The first projection optical systems **45** illuminate the anterior segment with infrared light and are also utilizable as index for measurement of corneal shape (kerato index).

The observation optical system **50** includes the reflection mirror **72**, the objective lens **14**, the dichroic mirror **53**, an imaging lens **51**, and a two-dimensional imaging element **52**. An output signal from the imaging element **52** is connected to the control part **80** and output to the monitor **85**. The observation optical system **50** is used for observation of a front image of the eye **E** and also used as a detecting optical system for detecting an alignment state of the main unit **100** with respect to the eye **E**.

An anterior segment image of the eye **E** illuminated by the first projection optical systems **45** is received by an imaging plane of the imaging element **52** via the reflection mirror **72**, the objective lens **14**, the dichroic mirror **53**, and the imaging lens **51**. Similarly, the alignment index images formed by the first projection optical systems **45** and the second projection optical systems **46** are detected by the imaging element **52**.

FIG. 3 is a block diagram showing electric and control systems contained in the main unit **100** in the first example. The control part **80** performs control of the whole device and arithmetic processing such as calculation of eye refractive values. The control part **80** is connected to the light source **11**, the imaging element **26**, the light source **31**, the imaging element **52**, the operation part **84** to be used for various settings, the monitor **85**, the deviation detecting unit **60**, the deviation compensating unit **70**, a memory **81**, and others. For activation of the detecting unit **60** and the compensating unit **70** by the control part **80**, a dedicated drive circuit (e.g., an LSI) may be used for speedup. Of course, they may be activated by software.

The control part **80** controls the monitor **85** to display an anterior segment image, a measurement result, and others on a screen in superimposing manner. The control part **80** further detects misalignment (alignment displacement) based on an imaging signal from the imaging element **52**.

FIG. 4 shows a ring image captured by the imaging element **26** during measurement. An output signal from the imaging element **26** is stored as image data (measurement image) in the memory **81**. Thereafter, the control part **80** detects an image position in each meridian direction based on the image stored in the memory **81**, and then performs elliptic approximation using a least square method and others. The control part **80** then determines a refraction error in each meridian direction from the approximated elliptic shape and, based on this determined error, measures eye refraction values; S (Spherical power), C (Cylinder power), and A (Astigmatic axial angle), and displays these values on the monitor **85**.

Operations of the device configured as above will be explained. An examiner holds the main unit **100** and instructs an examinee to look at the fixation target **32**, and then puts the test window **102** in front of the eye **E**. Accordingly, an anterior segment is imaged by the imaging element **52**. The monitor **85** displays an anterior segment image **F**, a ring image (Meyer ring image) **R** formed by the first projection optical systems **45**, and infinite index images **M** projected by the second projection optical systems **46** (see FIG. 3).

The control part **80** detects the alignment state with respect to the examinee's eye based on an imaging signal from the

imaging element **52**. In this case, the control part **80** determines misalignment in the X and Y directions by calculating a center position (a nearly corneal center) of the ring index **R**. In case the main unit **100** is displaced in the Z direction, the control part **80** determines misalignment in the Z direction by utilizing characteristics that the interval between the indexes **M** hardly changes but an image interval of the ring index **R** changes (for the details, refer to JP-A-6(1994)-46999). The control part **80** increases and decreases the number of indicators **G** based on an alignment detection result in the Z direction.

Herein, the examiner moves the main unit **100** in the X and Y directions so that the ring image **R** and a reticle mark **LT** become concentric. While referring to the indicators **G** changing based on the alignment detection result in the Z direction (or to adjust the ring image **R** so as to be thinnest), the examiner moves the main unit **100** in the Z direction.

In the case where auto-shot is thereafter activated, when the alignment state in the X, Y, and Z directions meets a permissible range, the control part **80** generates a trigger signal to start measurement. On the other hand, in the case where auto-shot is turned off, measurement is started upon pressure of a trigger switch provided in the operation part **84**.

When a trigger signal is output, the control part **80** turns on the light source **11** to project measurement index onto the fundus **Ef**. The control part **80** receives the reflection light through the imaging element **26** and detects the index image.

At that time, preliminary measurement is first conducted. Based on a result of this preliminary measurement, the light source **31** and the fixation target plate **32** are moved in the optical axis direction to apply a fogging to the eye **E**. Thereafter, main measurement on the eye **E** is performed. In this main measurement, images of a plurality of frames are captured and stored in the memory **81** for addition/accumulation processing of ring images, or multiple measurements.

In the above measurement, the control part **80** supplies power to the sensor **62** and generates a positional displacement signal based on an acceleration signal output from the sensor **62**. Upon input of the generated positional displacement signal, the control part **80** further outputs a deviation compensating signal to rotate the reflection mirror **72** in a direction to cancel the shift deviation of the main unit **100**.

FIGS. 5(a) to 5(c) show concrete examples to compensate hand shake during measurement in the first example; FIG. 5(a) shows a state before hand shake occurs, FIG. 5(b) shows a state immediately after shift displacement occurs due to hand shake, and FIG. 5(c) shows a state after the shift displacement is compensated. The control part **80** drives the deviation compensating unit **70** to bring a measurement region of the eye refractive power measuring optical system onto a pupil.

To compensate the hand shake, the optical axis **L1** is moved in an opposite direction to movement of the main unit **100** to cancel the displacement of the optical axis **L1** with respect to the center of the eye **E** (e.g., a pupil center **Pc**, or a corneal center) caused by the hand shake. It is to be noted that the displacement of the optical axis **L1** does not need to be cancelled completely as long as it is reduced.

For instance, when the main unit **100** is displaced downward by ΔY with respect to a visual line of the eye **E** due to hand shake of the examiner, a downward deviation amount ΔY is generated as a positional displacement signal. The control part **80** then drives the drive part **74** to rotate the reflection mirror **72** upward by $\Delta \theta$ to make the optical axis **L1** swing about the pupil center **Pc** as a swing point.

Measurement light from the light projecting optical system **10a** is compensated (deflected) by movement of the reflection

mirror **72** and then projected onto the fundus **Ef** via the pupil center **Pc**. Reflection light from the fundus is reflected by the reflection mirror **72** and made coaxial with the optical axis of the objective lens **14** and then directed toward the imaging element **26**. A measurement region by the ophthalmoscopic optical system **10** tracks a pupil of the eye **E**.

Fixation light of the target projecting optical system **30** is similarly compensated (deflected) by movement of the reflection mirror **72**, thereby inducing the eye **E** to make the visual line thereof coaxial with the optical axis **L1**. Specifically, the fixation direction of the eye **E** tracks the optical axis **L1**. Depending on a reaction speed of the eye **E**, the fixation direction may remain in a front direction. In the case where the target projecting optical system **30** is placed independent from the reflection mirror **72**, the eye **E** is fixed in the front direction.

Reflection light from the anterior segment is reflected by the reflection mirror **72**, and made coaxial with the optical axis of the objective lens **14**, and directed toward the imaging element **52**. An observation region by the observation optical system **50** tracks the anterior segment of the eye **E**.

Accordingly, even when the main unit **100** is moved due to hand shake of the examiner, measurement of refractive power, induction of fixation, and observation of anterior segment are enabled. In this case, it is particularly effective in projecting and receiving measurement light successively (e.g., obtaining measurement images of a plurality of frames) to measure refractive power. The above explanation is given to an example only on the control in the **Y** direction. When the same control is applied to the **X** direction, furthermore, a positional displacement deriving from hand shake can be compensated in the **X** and **Y** directions.

Since the displaced state of the optical axis **L1** from the reference position of the eye **E** (e.g., a corneal apex, or a pupil center) is detected and the compensating optical system is driven accordingly, even the hand-held ophthalmological device difficult to make alignment can perform stable measurement/examination.

As shown in FIG. 2, the deviation compensating optical system (**72**) is placed in an common optical path of the light projecting system and the light receiving system of the ophthalmoscopic optical system **10**, so that an incoming position and an outgoing position of light with respect to the eye are appropriately compensated, allowing accurate measurement (examination) to be performed. The deviation compensating optical system may also be placed in each of the light projecting system and the light receiving system.

As shown in FIG. 2, since the deviation compensating optical system is placed in the common optical path of the ophthalmoscopic optical system **10** and the observation optical system **50**, the deviation of the observation image is also compensated. This enables stable observation and measurement (examination).

In the above explanation, an optical reflection member (e.g., a reflection mirror, a reflection prism) is used as a compensating optical member to deflect ophthalmoscopic light to make displacement compensation. Accordingly, the light incident on the compensating optical member is caused to travel toward the eye **E**. Thus, disturbed light is avoided from entering the ophthalmoscopic light receiving element during movement of the optical member.

In this case, when a light deflection member (e.g., a prism) is provided in a position off the pupil conjugation position and this member is eccentrically rotated about the optical axis **L1**, the disturbed light is further appropriately removed.

Of course, even another compensating optical member may achieve fixed effects. In this case, it is preferably

arranged such that, when the main unit **100** and the eye **E** are displaced in position, the optical member is moved to maintain a conjugated relationship between the pupil of the eye **E** and the ring lens **24** in the **X** and **Y** directions. For example, the objective lens **14** placed in the common optical path of the light projecting optical path and the light receiving optical path may be moved vertically with respect to the optical axis. Of course, an optical member (e.g., a concave lens) special for optical axis displacement may be placed. As another alternative, a plurality of optical members may be provided.

Furthermore, the ring lens **24** and the imaging element **26** may be moved in a vertical direction with respect to the optical axis of the lens **22** (the light source **11** may be moved in sync therewith). The imaging element **26** may be used as a deviation detecting system. For instance, based on an imaging signal from the imaging element **26**, the compensating optical system is moved to make the center of a ring image on the imaging element **26** coincide with the optical axis.

The above explanation uses the sensor (the detecting unit **60**) configured to detect the movement of the main unit **100**. As an alternative, the imaging element configured to image the eye **E** may be used as a deviation detecting sensor. For instance, the control part **80** detects a deviation from an imaging result of the imaging element **52**, and feedback drives the compensating optical system so that the corneal center (or a pupil center) falls within a permissible range from a certain alignment reference position (e.g., an intersection point between the imaging plane and the optical axis **L1**). According to the present technique, even when the eye **E** moves with respect to the main unit **100** in the driving range of the compensating optical system, the position displacement is compensated.

In the above explanation, hand shake detection and compensating operation may be started at the time when the misalignment falls within a certain permissible range (e.g., wider than an alignment completion range). As an alternative, the hand shake compensating operation may be started in response to an output signal from the operation part **84**.

When the misalignment is to be detected from output from the imaging element **52**, a deviation amount detected by the detecting unit **60** with respect to an actual displacement amount may be offset (compensated). At the time of compensating the hand shake, alignment detection and result output may be stopped.

The above explanation exemplifies the auto-refractometer. However, the present invention is also applicable to any other ophthalmological devices. For instance, the invention is also applicable to a hand-held type fundus photographing device (e.g., a fundus camera, an ophthalmic OCT). In this case, the displacement between the imaging optical axis and the eye **E** due to hand shake is detected and the compensating optical system is driven based on a detection signal thereof.

In the case of the device of imaging a fundus, for example, a control part may detect a deviation of a fundus image captured by an imaging element for fundus imaging (e.g., for alignment) provided in a fundus observation optical system and feedback drive a compensating optical system to compensate the deviation. Of course, an accelerator sensor and others may also be provided.

The present invention is applicable to not only the hand-held ophthalmological device but also a floor-standing ophthalmological device in case it is installed in an unstable place.

In the above explanation, the hand-held ophthalmological device configured to project measurement light onto a fundus and receive reflection light therefrom to measure eye characteristics (e.g., eye refractive power, ocular axial length) uses

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the sensor configured to detect a deviation applied to the main unit **100**. However, the invention is not limited thereto.

For instance, a light deflection member (e.g., a prism, a mirror, etc.) may be provided in an optical path of a measuring optical system and driven to change a passage region of the measurement light on an anterior segment as needed to measure a deviation. The light deflection member is placed for example in a position off a conjugated position of the measuring optical system with a pupil. To be specific, a prism is eccentrically rotated about the optical axis **L1** to eccentrically rotate the passage region. Further, the mirror is reciprocated up and down to move the passage region up and down.

With the above manner, the passage region of the measurement light is rapidly moved with respect to the anterior segment, thereby cancelling out the displacement of the measurement light due to hand shake. Accordingly, the measurement is easily performed by receiving the measurement light at the time of cancelling.

Second Example

A second example of the present embodiment will be explained below. FIG. **6** is an external side view showing one example of a hand-held ophthalmological device of the second example. The following explanation exemplifies an eye refractive power measuring device (an auto-refractometer). In the following explanation, a positional relationship between an examinee's eye and the device is defined assuming that a back and forth direction is a Z direction, a right and left direction is an X direction, and an up and down direction is a Y direction while the examinee's eye and the device face each other.

The main unit **100** is provided with the optical unit **1** and a compensating unit **110**. The optical unit **1** includes the ophthalmoscopic optical system **10** to project an ophthalmoscopic light beam onto the examinee's eye and receive reflection light therefrom. The test window **102** is positioned in the side of the main unit **100** that will face the eye **E**. The operation part **84** and the monitor **85** are positioned in the side of the main unit **100** that will face an examiner. Furthermore, the main unit **100** contains an electric system for control and arithmetic processing. The examiner faces the examinee and holds the main unit **100**. While looking at the monitor **85**, the examiner makes alignment of the main unit **100** with respect to the eye **E**.

The compensating unit **110** is provided to compensate a relative positional displacement between the ophthalmoscopic optical axis **L1** of the ophthalmoscopic optical system **10** and the eye **E**. The compensating unit **110** is provided with for example the objective lens **14** and a drive part **111**, which serve as the compensating optical system. The objective lens **14** is placed movably on a plane vertical to the optical axis of the objective lens **14**. The drive part **111** is provided to shift (move) the objective lens **14** in a vertical direction (a direction perpendicular to the measurement optical axis **L1**) with respect to the optical axis of the objective lens **14**. The objective lens **14** is placed in one of the optical paths of the ophthalmoscopic optical system **10** and is used as a part of the ophthalmoscopic optical system **10**.

In FIG. **6**, the optical axis **L1** of the ophthalmoscopic optical system **10** is deflected toward the eye **E** by the reflection mirror **72**. The reference position of the objective lens **14** is a position in which the optical axis **L1** formed by a lens system (e.g., a lens **12**) other than the objective lens **14** of the ophthalmoscopic optical system **10** and the optical axis of the objective lens **14** become coaxial. In the present example, the optical axis **L1** of the ophthalmoscopic optical system **10** is

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deflected toward the eye **E** by the reflection mirror **72**, but it is not limited to this configuration. For instance, it may be arranged so that the optical axis **L1** of the ophthalmoscopic optical system **10** directly goes to the eye **E**.

When the objective lens **14** is shifted by driving of the drive part **111**, the optical axis **L1** is deflected, thereby adjusting the optical axis **L1** to pass through the pupil center **Pc** (see FIG. **10**).

FIG. **7** is an optical arrangement diagram of the optical systems contained in the main unit **100** seen from front in the second example. The optical unit **1** is provided with the ophthalmoscopic optical system **10**, the target projecting optical system **30** to project a fixation target to the eye **E**, the observation optical system **50** to allow observation of the eye **E**, and a front projection optical system **120** to form a luminescent spot (a bright spot) at a corneal apex of the eye **E**.

In FIG. **7**, the ophthalmoscopic optical system **10** is an optical system to objectively measure eye refractive power. The ophthalmoscopic optical system **10** projects a measurement index to an examinee's eye and receive reflection light from the examinee's eye acquired by the measurement index. Based on the light receiving signal, the refractive power of the eye **E** is measured.

To be specific, the ophthalmoscopic optical system **10** consists of a light projecting optical system **10a** to project a measurement index (a spot index) onto the fundus **Ef** through the pupil center **Pc** and a light receiving optical system **10b** to extract fundus reflection light as a ring image through the periphery of a pupil and form (receive) a ring-shaped reflection image on the imaging element **26**.

The light projecting optical system **10a** includes the light source **11**, the relay lens **12**, the hole mirror **13**, the objective lens **14**, and the reflection mirror **72**, which are arranged on the optical axis **L1**. The light source **11** is placed in a nearly conjugated position with the fundus **Ef** and an aperture of the hole mirror **13** is placed in a nearly conjugated position with the pupil of the eye **E**.

Measurement light emitted from the light source **11** is reflected by the reflection mirror **72** via the relay lens **12**, the hole mirror **13**, the dichroic mirror **35**, the dichroic mirror **53**, and the objective lens **14**. The measurement light reflected by the reflection mirror **72** passes through the pupil center **Pc**, projecting a spot-shaped beam onto the fundus **Ef**.

The light receiving optical system **10b** includes the reflection mirror **72**, the objective lens **14**, the hole mirror **13**, the collimator lens **22**, the ring lens **24**, and the imaging element (e.g., a two-dimensional imaging element such as a CCD and a CMOS) **26**. The imaging element **26** is placed in a nearly conjugated position with the fundus **Ef** via the objective lens **14**, the lens **22**, and the ring lens **24**. The ring lens **24** consists of a lens part formed with an annular, cylindrical lens and a light shielding part having a ring aperture equal in size to the lens part. The ring lens **24** is placed in a nearly conjugated position with the pupil of the eye **E** with respect to the objective lens **14**. An output signal from the imaging element **26** is connected to the control part **80**.

The fundus reflection light, which is formed by projection to the fundus **Ef** by the light projecting optical system **10a** and reflection from the fundus **Ef**, is reflected again by the reflection mirror **72**, and reflected again by the reflection plane of the hole mirror **13** via the objective lens **14**, the dichroic mirror **53**, the dichroic mirror **35**, and then is collimated into a nearly parallel beam (in a case of an emmetropic eye) by the collimator lens **22**. The ring-shaped beam is extracted by the ring lens **24** and received as a ring image by the imaging element **26**.

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Instead of the above configurations, various methods may be adopted; e.g., a method of projecting a ring index onto the fundus Ef through the peripheral part of a pupil and extracting reflection light through the pupil center so that a ring image is received by the imaging element, a phase difference method performed by projecting slit light onto a fundus, and others. Furthermore, a configuration to extract an intermittent ring image, not a continuous ring image, may be adopted. Another configuration to extract a fundus reflection image consisting of point images arranged in a nearly ring form may also be adopted.

The target projecting optical system 30 includes the light source 31 that emits visible light, the fixation target 32 on which a scenic/animal image or others is painted, the light projecting lens 33, a dichroic mirror 125, the dichroic mirror 35, the objective lens 14, and the reflection mirror 72. In addition to the above configuration, as a fixation target, a point light source such as LED and a display such as a liquid crystal display, and others are used. Furthermore, a plurality of fixation targets may be placed two-dimensionally.

The fixation target 32 illuminated by the light source 31 is projected onto the fundus Ef via the light projecting lens 33, the dichroic mirror 125, the dichroic mirror 35, the dichroic mirror 53, the objective lens 14, and the reflection mirror 72. Accordingly, the eye E is induced to hold fixation. The light source 31 and the fixation target 32 are moved in the optical axis direction to apply a fogging to the eye E.

The front projection optical system 120 is provided with a light source 121 that emits near infrared light, a light projecting lens 122, the dichroic mirror 35, the objective lens 14, and the reflection mirror 72. The light source 121 used herein is for example an LED (Light Emitting Diode) light source.

Light emitted from the light source 121 is made into nearly parallel light by the objective lens 14 via the light projecting lens 122, the dichroic mirror 125, the dichroic mirror 35, and the dichroic mirror 53, and then reflected by the reflection mirror 72. Light reflected by the reflection mirror 72 is projected as an infinite index onto a cornea Ec. The front projection optical system 120 projects an index onto the eye E from front, thereby forming a corneal luminescent spot (an index image) at an apex of the cornea Ec. This corneal luminescent spot is used for detection of the relative positional displacement between the ophthalmoscopic optical axis L1 and the eye E (the details will be described later).

Between the light projecting lens 122 and the dichroic mirror 35, the dichroic mirror 125 that reflects visible light and transmits infrared light is placed to deliver the light from the light source 31 toward the eye E. Between the objective lens 14 and the hole mirror 13, the dichroic mirror 35 serving as a beam splitter is placed to deliver the light from the light source 31 and the light source 121 to the eye E. The dichroic mirror 35 has wavelength characteristics that reflect light of the light source 121 and the light source 31 and transmit light of the light source 11. Between the dichroic mirror 35 and the objective lens 14, furthermore, the dichroic mirror 53 serving as a beam splitter is placed to direct the light from an anterior segment to the observation optical system 50. The mirror 53 has wavelength characteristics that transmit measurement light and reflect observation light. By those beam splitters, the measurement optical axis, the fixation optical axis, and the observation optical axis are made coaxial with the same optical axis (L1).

On the outside of the test window 102, the first projection optical systems 45 to project ring-shaped finite index images onto the cornea Ec are arranged concentrically with the optical axis L1. Furthermore, the second projection optical systems 46 to project infinite index images onto the cornea Ec are

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arranged left-right symmetric (up-down symmetric for convenience in the figure) with respect to the optical axis L1. The first projection optical systems 45 illuminate the anterior segment with infrared light and are also utilizable as index for measurement of corneal shape (kerato index).

The observation optical system 50 includes the reflection mirror 72, the objective lens 14, the dichroic mirror 53, the imaging lens 51, and the two-dimensional imaging element 52. An output signal from the imaging element 52 is connected to the control part 80 and output to the monitor 85. The observation optical system 50 is used for observation of a front image of the eye E and also used as a detecting optical system for detecting an alignment state of the main unit 100 with respect to the eye E.

An anterior segment image of the eye E is received by an imaging plane of the imaging element 52 via the reflection mirror 72, the objective lens 14, the dichroic mirror 53, the dichroic mirror 54, and the imaging lens 51. Similarly, the alignment index images formed by the first projection optical systems 45 and the second projection optical systems 46 are detected by the imaging element 52. Between the dichroic mirror 53 and the imaging lens 51, a dichroic mirror 54 serving as a beam splitter is placed to direct the corneal reflection light formed by the light source 121, of the light from the anterior segment, to an XY alignment detecting optical system 150 (hereinafter, simply referred to as a detecting optical system 150). The mirror 54 has wavelength characteristics that transmit light of the first projection optical systems 45 and the second projection optical systems 46 and reflect light of the front projection optical system 120.

The detecting optical system 150 is provided to rapidly detect the relative positional displacement between the ophthalmoscopic optical axis L1 of the ophthalmoscopic optical system 10 and the eye E. The detecting optical system 150 includes the reflection mirror 72, the objective lens 14, the dichroic mirror 53, the dichroic mirror 54, a light receiving lens 55, and an optical position sensor (PSD: Position Sensitive Detector) 56.

For instance, the detecting optical system 150 is used to detect the positional displacement based on a corneal luminescent spot formed by the front projection optical system 120 and detected by the PSD 56. An output signal from the PSD 56 is connected to the control part 80 and output to the monitor 85.

The PSD is a semiconductor position detecting device, which is a photosensor capable of detecting the position of a spot light. An output signal from the PSD is an output signal representing positional data of the corneal luminescent spot itself and thus enables prompt acquisition of positional data of the corneal luminescent spot.

The PSD enables enhanced speed of the processing from light reception to displacement detection as compared with a two-dimensional imaging element such as CCD. Accordingly, since a severe positional displacement due to hand shake or the like is detected by use of the PSD, the displacement can be compensated at high speed.

The corneal center luminescent spot (the index image) projected by the front projection optical system 120 is received by the light receiving plane of the PSD 56 via the reflection mirror 72, the objective lens 14, the dichroic mirror 53, the dichroic mirror 54, and the imaging lens 55.

FIG. 8 is a block diagram showing electric and control systems contained in the main unit 100 in the second example. The control part 80 performs control of the whole device and arithmetic processing such as calculation of eye refractive values. The control part 80 is connected to the light source 11, the imaging element 26, the light source 31, the

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light source 121, the imaging element 52, the PSD 56, the operation part 84 to be used for various settings, the monitor 85, the compensating unit 110, the memory 81, and others. For activation of the compensating unit 110 by the control part 80, a dedicated drive circuit (e.g., an LSI) may be used for speedup. Of course, they may be activated by software.

The control part 80 controls the monitor 85 to display an anterior segment image, a measurement result, and others on a screen in superimposing manner. The control part 80 further detects a misalignment amount and a hand shake amount based on an imaging signal from the PDS 56. In the present example, misalignment caused by an examinee and hand shake caused by an examiner are detected by the PSD 56. As an alternative, the misalignment detection and the hand shake detection may be performed based on the imaging signal from the imaging element 52.

FIG. 4 shows a ring image captured by the imaging element 26 during measurement. An output signal from the imaging element 26 is stored as image data (measurement image) in the memory 81. Thereafter, the control part 80 detects an image position in each meridian direction based on the image stored in the memory 81, and then performs elliptic approximation using a least square method and others. The control part 80 then determines a refraction error in each meridian direction from the approximated elliptic shape and, based on this determined error, measures eye refraction values; S (Spherical power), C (Cylinder power), and A (Astigmatic axial angle), and displays these values on the monitor 85.

Operations of the device configured as above will be explained. An examiner holds the main unit 100 and instructs an examiner to look at the fixation target 32, and then puts the test window 102 in front of the eye E. Accordingly, an anterior segment is imaged by the imaging element 52. The monitor 85 displays an anterior segment image. FIG. 9 shows one example of a display screen of the monitor 85. On the monitor 85, there are displayed the anterior segment image F, a ring image (Meyer ring image) R formed by the first projection optical systems 45, infinite index images M projected by the second projection optical systems 46, and a corneal center index (an infinite corneal center luminescent spot) C projected by the front projection optical system 120.

The control part 80 detects a relative positional displacement between the eye E and the optical axis L1 based on a light receiving signal from the PSD 56. The relative positional displacement includes at least one of a displacement caused when the main unit 100 is moved with respect to the eye E (e.g., hand shake), a displacement caused when the eye E is moved with respect to the main unit 100 (e.g., involuntary eye movement), and a displacement caused before completion of alignment with respect to the eye E.

For instance, based on the light receiving signal from the PSD 56, the control part 80 detects the relative displacement between the optical axis L1 of the ophthalmoscopic optical system 10 and the examinee's eye in a direction perpendicular to the optical axis L1 of the ophthalmoscopic optical system 10. The control part 80 calculates the corneal center luminescent spot C to determine the positional displacement in the X and Y directions of the optical axis L1 with respect to the eye E. The control part 80 utilizes the characteristics that when the main unit 100 is displaced in the Z direction, the interval between the indexes M hardly changes but the image interval of the ring index R changes (for the details, refer to JP-A-6 (1994)-46999). The control part 80 increases and decreases the number of indicators G based on an alignment detection result in the Z direction.

Herein, the examiner moves (shifts) the main unit 100 in the X and Y directions so that the corneal center luminescent

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spot C falls within the reticle mark LT. The reticle mark LT in the present example electronically represents the reference position set as a position where the corneal apex position and the optical axis L1 of the device coincide with each other. While referring to the indicators G changing based on the alignment detection result in the Z direction (or to adjust the ring image R so as to be thinnest), the examiner moves the main unit 100 in the Z direction.

In the case where auto-shot is thereafter activated, when the alignment state in the X, Y, and Z directions falls within a permissible range, the control part 80 generates a trigger signal to start measurement. On the other hand, in the case where auto-shot is turned off, measurement is started upon pressure of a trigger switch provided in the operation part 84.

When the trigger signal is output, the control part 80 turns on the light source 11 to project measurement index to the fundus Ef. The control part 80 further turns off the first projection optical systems 45, the second projection optical systems 46, the front projection optical system 120 (the light source 121). Specifically, to prevent overlapping of the measurement light with another light from influencing a measurement result or an image to be observed, the control part 80 turns on the light source 11 at the time when the first projection optical systems 45 to the front projection optical system 120 are turned off. The control part 80 turns off the light source 11 at the time when the first projection optical systems 45 to the front projection optical system 120 are turned on. Thus, the control part 80 receives the reflection light by the imaging element 26 and detects the index image.

At that time, preliminary measurement is first conducted. Based on a result of this preliminary measurement, the light source 31 and the fixation target plate 32 are moved in the optical axis direction to apply a fogging to the eye E. Thereafter, main measurement on the eye E is performed. In this main measurement, images of a plurality of frames are captured and stored in the memory 81 for addition/accumulation processing of ring images, or multiple measurements.

At that time, when measurement is performed more than one time, the control part 80 makes detection of an alignment state and detection of hand shake by the first projection optical systems 45 to the front projection optical system 120 every time one frame image is acquired.

For instance, when the control part 80 starts measurement and then acquires one frame image, the control part 80 turns off the light source 11 and turns on the first projection optical systems 45 to the front projection optical system 120. The control part 80 drives the drive part 111 based on the light receiving signal from the PSD 56 to compensate the positional displacement. After completion of displacement compensation, the control part 80 turns on the light source 11 at the time of turning off the first projection optical systems 45 to the front projection optical system 120 and acquires a next one frame image. During measurement, specifically, the light of the light source 11 and the light of another optical system (e.g., the first projection optical systems 45, the second projection optical systems 46, and the front projection optical system 120) are alternately blinked at high speed to perform measurement. Respective lights of the light source 11, the first projection optical systems 45, the second projection optical systems 46, and the front projection optical system 120 may be fast blinked in turn.

An explanation is given to the hand shake compensation using the front projection optical system. The device in the present example compensates the shift displacement in the X and Y directions of the main unit 100 caused by hand shake.

The control part 80 detects, through the PSD 56, the shift displacement (displacement amount and displacement direc-

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tion) between the corneal luminescent spot C formed on the cornea Ec and the reticle LT. At that time, a displacement amount until the corneal center luminescent spot C comes to enter a predetermined permissible range of the previously set reticle mark LT (the reference position) is detected as shift displacement. The control part 80 drives the drive part 111 based on the detected shift displacement to shift the objective lens 14, thereby compensating the shift displacement caused by the hand shake. Specifically, the control part 80 outputs a displacement compensating signal to shift the objective lens 14 in a direction to cancel out the shift deviation of the main unit 100.

The aforementioned predetermined permissible range is set for example to a range in which a positional displacement can be compensated by driving of the compensating unit 110 (e.g., within a range of 1.0 mm displacement in each direction).

FIGS. 10(a) to 10(c) show concrete examples to compensate hand shake during measurement in the second example; FIG. 10(a) shows a state before hand shake occurs, FIG. 10(b) shows immediately after a shift displacement is caused by hand shake, and FIG. 10(c) shows a state after the shift displacement is compensated. The control part 80 drives the deviation compensating unit 110 to bring the measurement region formed by the eye refractive power measuring optical system to a position on the pupil.

To compensate the hand shake, the control part 80 controls driving of the drive part 111, and moves the optical axis L1 in an opposite direction to movement of the main unit 100 in order to cancel the displacement of the optical axis L1 with respect to the center of the eye E (e.g., the pupil center Pc or the corneal center) caused by the hand shake. It is to be noted that the displacement of the optical axis L1 does not need to be cancelled completely as long as it is reduced.

For instance, when the main unit 100 is displaced downward by ΔY with respect to a visual line of the eye E due to hand shake of the examiner, a downward displacement ΔY is detected. The control part 80 then drives the drive part 111 to shift (move) the lens 14 by ΔS in a direction indicated by an arrow A to compensate the shift displacement ΔY .

Measurement light from the light projecting optical system 10a is compensated (deflected) by movement of the objective lens 14 and then projected onto the fundus Ef via the pupil center Pc. The reflection light from the fundus is compensated by the objective lens and made coaxial with the optical axis of the objective lens 14, and directed toward the imaging element 26. The measurement region by the ophthalmoscopic optical system 10 tracks the corneal apex of the eye E.

Fixation light of the target projecting optical system 30 is similarly compensated (deflected) by movement of the objective lens 14, thereby inducing the eye E to hold fixation on the test window 102, so that the visual line of the eye E becomes coaxial with the optical axis L1. Depending on a reaction speed of the eye E, the fixation direction may remain a front direction. In the case where the target projecting optical system 30 is placed independent from the objective lens 14, the eye E is fixed in the front direction.

When the objective lens 14 is moved from the reference position, the reflection light formed by the first projection optical systems 45 and the second projection optical systems 46 and reflected from the anterior segment is deflected by the objective lens 14, and made coaxial with the optical axis of the lens 51 placed downstream of the objective lens 14. This light is received by the imaging element 52.

When the objective lens 14 is moved from the reference position, the reflection light formed by the front projection optical system 120 and reflected from the anterior segment is

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deflected by the objective lens 14 to become coaxial with the optical axis of the lens 55 disposed downstream of the objective lens 14. This light is received by the PSD 56.

The above configuration enables measurement of eye refractive power, induction of fixation, and observation of anterior segment even when the main unit 100 is relatively moved due to hand shake of the examiner or movement of the examinee. In this case, it is particularly effective in continuously projecting and receiving measurement light (e.g., obtaining measurement images of a plurality of frames) to measure refractive power. The above explanation is given to an example only on the control in the Y direction. When the same control is applied to the X direction, furthermore, the positional displacement deriving from hand shake can be compensated in the X and Y directions.

Since the displacement of the optical axis L1 from the eye E is detected and the compensating optical system is driven accordingly, even the hand-held ophthalmological device difficult to make alignment can perform stable measurement/examination.

The above compensation is also effective in compensating a displacement caused when the eye E moves with respect to the main unit 100 (e.g., involuntary eye movement) and a displacement caused before completion of alignment with respect to the eye E.

In the case of the hand-held ophthalmological device, including no alignment movement mechanism (e.g., a motor mechanism to move the main unit in three dimensional directions, and others) of the floor-standing ophthalmological device, it is difficult to move the main unit to an alignment completion position. When continuous measurement is performed after completion of alignment, the eye moves by involuntary eye movement and thus re-adjustment is needed. Since it is normally necessary to measure both eyes and make alignment for each eye, alignment adjustment takes time.

According to the above configuration, the examiner has only to move the main unit 100 to a range where the positional displacement can be compensated by driving of the compensating unit 110. This largely improves an alignment operation of the hand-held ophthalmological device.

As shown in FIG. 7, the deviation compensating unit 110 is placed in a common optical path of the light projecting system and the light receiving system of the ophthalmoscopic optical system 10, so that an incoming position and an outgoing position of light with respect to the eye are appropriately compensated, allowing accurate measurement (examination) to be performed. The deviation compensating optical system may also be placed in each of the light projecting system and the light receiving system.

As shown in FIG. 7, the deviation compensating optical system is placed in the common optical path of the ophthalmoscopic optical system 10 and the observation optical system 50, a deviation of the observation image is also compensated. This enables stable observation and measurement (examination).

In the present example, the hand shake compensation in the Z direction may be further performed. This hand shake compensation in the Z direction may be conducted by a configuration to adjust focus using an optical member or by compensating a measurement result from the displacement amount in the Z direction. When the measurement result is to be compensated from the displacement amount in the Z direction, for example, the measurement result is compensated by use of a measurement result compensating table of compensation amounts set according to displacement amounts in the Z direction. The measurement result compensating table may be created in advance by calculating compensation values of

measurement results according to displacement amounts in the Z direction, and stored in the memory **81**.

Of course, even another compensating optical member can provide fixed effects. For instance, a light reflection member (e.g., a reflection mirror or a reflection prism) may be used as a compensating optical system to deflect ophthalmoscopic light in order to compensate the displacement. Of course, a special optical member (e.g., a concave lens) for optical axis displacement may be disposed. A plurality of optical members may be adopted.

In the present example, if the corneal center luminescent spot is not detected, the misalignment may be compensated to a position where the corneal center luminescent spot is detectable by use of the ring image R projected by the first projection optical systems. An alternative is to detect misalignment in the X and Y directions by the ring image R projected by the first projection optical systems and detect a shift displacement due to hand shake by the corneal center luminescent spot.

If the displacement amount between the optical axis L1 and the eye E exceeds a certain permissible range (e.g., a range that is wider than an alignment completion range and allows compensation by lens movement), the control part **80** may cause the monitor **85** to display an indication to prompt movement of the main unit **100**. For instance, an arrow or other indication for helping alignment may be displayed on the monitor **85**. In this case, the control part **80** may stop compensation control using the lens.

In the present example, the control part **80** may detect a rough positional displacement between the eye E and the optical axis L1 based on an imaging signal from the imaging element **52**, and detect a fine positional displacement between the eye E and the optical axis L1 based on the light receiving signal from the PSD **56**.

In the present example, the hand shake detection and the compensating operation may be started at the time when the misalignment falls within a certain permissible range (e.g., wider than the alignment completion range). As another alternative, the output signal from the operating part **84** may trigger the hand shake compensating operation. When the objective lens **14** reaches a limit of a movable range, the control part **80** stops driving of the deviation compensating unit **110**, returns the objective lens **14** to the reference position, and restarts the compensating operation. In this case, for example, the state where the test window **102** is placed in front of the eye E is detected based on the output signal from the PSD **56**, and the compensation is started.

In the present example, the configuration to detect the positional displacement is configured to optically detect a relative positional displacement between the optical axis L1 of the ophthalmoscopic optical system **10** and the anterior segment of the examinee's eye by use of the detecting optical system **120** (PSD **56**). However, the present invention is not limited thereto. It may be arranged to detect the relative positional displacement by utilizing both the detection result of the accelerator sensor and the detection result of the PSD to thereby compensate the positional displacement. For instance, the device in the second example is additionally provided with an accelerator sensor. The control part **80** controls the drive mechanism **111** based on a detection signal from the accelerator sensor to compensate the hand shake and controls the drive mechanism **111** based on a light receiving signal from the PSD to compensate the positional displacement caused when the eye moves.

In the present example, the control part **80** may compensate the measurement result of eye optical characteristics of the examinee's eye according to at least the position of the lens.

For instance, the measurement results may be compensated based on the misalignment amount in the X and Y directions, the focus displacement amount (displacement amount in the Z direction), and the position of the objective lens **14** for displacement compensation. This is because the measurement result may vary if the objective lens **14** is deviated with respect to another optical member of the ophthalmoscopic optical system **10**. In the hand-held ophthalmic measuring device (auto-refractometer (keratometer)), preferably, an anterior segment image, a pupil image, and others are stored together with the measurement result in the memory **81** during measurement. A positional relationship between the device and the eye E is detected based on such an image and is used for compensation of the measurement result (for example, astigmatic axial angle). In a case of performing the examination on a bedridden examinee who takes uneasy posture, for example, the control part **80** detects a tilted state between the device and the eye E by using an image stored in the memory **81**, and compensates the measurement result based on the tilted state.

In compensating the measurement result, for example, the measurement result compensating table is used for compensation of the measurement result. For the measurement result compensating table, compensating amounts according to positions of the objective lens **14** on a plane vertical to the optical axis of the objective lens **14** are determined in advance by experiments or simulation. Each compensating value of the measurement result corresponding to each position is calculated in advance to create the table.

In the present embodiment, when the light source **11** is turned on, the first projection optical systems **45**, the second projection optical systems **46**, and the front projection optical system **120** are turned off. The invention is not limited thereto. For instance, the first projection optical systems and the second projection optical systems may be held turned-on even during turn-on of the light source **11**. In this case, it is more preferable to use light sources different in wavelength to prevent interference of light between the light sources.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. **1** is an external side view showing one example of a hand-held ophthalmological device in a first embodiment;

FIG. **2** is an optical arrangement diagram of optical systems contained in a main unit seen from front in a first example;

FIG. **3** is a block diagram showing electric and control systems contained in the main unit of the first example;

FIG. **4** is a ring image captured by an imaging element during measurement;

FIGS. **5(a)** to **5(c)** show concrete examples to compensate hand shake during measurement in the first example;

FIG. **6** is an external side view showing one example of a hand-held ophthalmological device of a second example;

FIG. **7** is an optical arrangement diagram of optical systems contained in a main unit seen from front in the second example;

FIG. **8** is a block diagram showing electric and control systems contained in the main unit of the second example;

FIG. **9** shows one example of a display screen of a monitor of the second example; and

FIGS. **10(a)** to **10(c)** are concrete examples to compensate hand shake during measurement in the second example.

REFERENCE SIGNS LIST

10 Ophthalmoscopic optical system
14 Objective lens

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45 First projection optical system
 46 Second projection optical system
 50 Observation optical system
 52 Imaging element
 56 PSD
 60 Deviation detecting unit
 70 Deviation compensating unit
 72 Reflection mirror
 74 Drive part
 100 Main unit
 110 Compensating unit
 111 Drive part
 120 Front projection optical system
 150 XY alignment detecting optical system

The invention claimed is:

1. A hand-held ophthalmological device including:
 - a main unit having an ophthalmoscopic optical system configured to project ophthalmoscopic light to an examinee's eye and receive reflection light therefrom to examine or measure an examinee's eye;
 - a detector placed in the main unit and configured to detect a relative deviation between an optical axis of the ophthalmoscopic optical system and the examinee's eye;
 - a deviation compensating optical system placed as a part of the ophthalmoscopic optical system and configured to compensate the deviation; and
 - a drive part configured to drive the deviation compensating optical system based on output from the detector.
2. The hand-held ophthalmological device according to claim 1, wherein the detector includes a shift deviation detector configured to detect a shift deviation which is applied to the main unit.
3. The hand-held ophthalmological device according to claim 1, wherein the detector receives reflection light from a cornea of the examinee's eye to detect the relative deviation between the optical axis of the ophthalmoscopic optical system and the examinee's eye.
4. The hand-held ophthalmological device according to claim 1, wherein the deviation compensating optical system is placed in a common optical path of a light projecting system and a light receiving system of the ophthalmoscopic optical system or is placed in each of the light projecting system and the light receiving system of the ophthalmoscopic optical system.
5. The hand-held ophthalmological device according to claim 1, wherein
 - the main unit further includes an imaging optical system having an imaging element configured to capture a front image of the examinee's eye, and
 - the deviation compensating optical system is placed in a common optical path of the ophthalmoscopic optical system and the imaging optical system.
6. The hand-held ophthalmological device according to claim 5, wherein the deviation compensating optical system is a light deflection member configured to deflect the ophthalmoscopic light.
7. The hand-held ophthalmological device according to claim 6, wherein the light deflection member is a light reflection member configured to reflect the ophthalmoscopic light toward the examinee's eye.
8. The hand-held ophthalmological device according to claim 7, wherein the ophthalmoscopic optical system is an eye refractive power measuring optical system configured to project a measurement index onto a fundus of the examinee's eye, and receive reflection light from the fundus to measure eye refractive power.

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9. The hand-held ophthalmological device according to claim 8, wherein the drive part drives the deviation compensating optical system to bring a measurement region by the eye refractive power measuring optical system onto a pupil of the examinee's eye.
10. The hand-held ophthalmological device according to claim 6, wherein the light deflection member is a lens that transmits the ophthalmoscopic light toward the examinee's eye.
11. The hand-held ophthalmological device according to claim 10, wherein
 - the ophthalmoscopic optical system is an eye refractive power measuring optical system configured to project a measurement index onto a fundus of the examinee's eye, and receive reflection light from the fundus to measure eye refractive power.
12. The hand-held ophthalmological device according to claim 11, wherein the drive part drives the deviation compensating optical system to bring a measurement region by the eye refractive power measuring optical system onto a pupil of the examinee's eye.
13. The hand-held ophthalmological device according to claim 1, wherein
 - the detector is an optical position sensor, and
 - the relative deviation between the optical axis and the examinee's eye is detected based on a light receiving signal from the optical position sensor.
14. A hand-held ophthalmological device including:
 - an ophthalmoscopic optical system including a light projecting optical system configured to project a measurement index onto an examinee's eye and a light receiving optical system configured to receive reflection light from the examinee's eye through a light receiving element, the reflection light being obtained by the measurement index, the ophthalmoscopic optical system being configured to measure eye optical characteristics of the examinee's eye based on an output signal from the light receiving element;
 - a detector configured to detect a relative positional displacement between an optical axis of the ophthalmoscopic optical system and an anterior segment of the examinee's eye;
 - a compensating unit including a drive part and being configured to move a lens placed in an optical path of the ophthalmoscopic optical system in a vertical direction with respect to an optical axis of the lens to compensate the positional displacement; and
 - a controller configured to control the drive part based on an output from the detector.
15. The hand-held ophthalmological device according to claim 14, wherein the detector optically detects the relative positional displacement between the optical axis of the ophthalmoscopic optical system and the anterior segment of the examinee's eye.
16. The hand-held ophthalmological device according to claim 15, further including a projection optical system configured to form an index image at a nearly corneal apex of the examinee's eye,
 - wherein the detector is an optical position sensor configured to detect a position of the reflection light obtained by the index image, and
 - the detector is arranged to detect a relative displacement between an optical axis of the ophthalmoscopic optical system and the examinee's eye in a direction perpendicular to the optical axis of the ophthalmoscopic optical system based on a light receiving signal from the optical position sensor.

17. The hand-held ophthalmological device according to claim 14, further including a computing unit configured to compensate a measurement result of eye optical characteristics of the examinee's eye according to a position of the lens.

18. The hand-held ophthalmological device according to claim 14, wherein the detector detects at least a relative deviation between the optical axis of the ophthalmoscopic optical system and the anterior segment of the examinee's eye.

19. The hand-held ophthalmological device according to claim 14, wherein the lens is an objective lens of the ophthalmoscopic optical system.

20. The hand-held ophthalmological device according to claim 14, wherein when the positional displacement exceeds a predetermined permissible range, the lens is returned to a reference position.

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